

**THE EFFECT OF INCREASING FOOT RIGIDITY ON  
MAXIMAL CYCLING POWER THROUGH THE USE OF  
CYCLING SPECIFIC ORTHOTICS**

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## Abstract

The purpose of this study was to investigate the performance enhancing capabilities of cycling specific orthotics during maximal cycling and how they relate to subject specific foot morphology and function. Twelve recreational cyclists took part in the study: eight male (age,  $38 \pm 8$  yr; height,  $180.41 \pm 3.55$  cm; body mass,  $80.90 \pm 6.50$  kg) and four female (age,  $35.92 \pm 20.82$  yr; height,  $176.57 \pm 1.94$  cm; body mass,  $77.20 \pm 2.05$  kg). Navicular height measurements were taken in weight bearing and non-weight bearing conditions to describe foot mobility. Subjects performed 2 maximal sprints (4s) on an isokinetic cycling ergometer at a cadence of 120rpm separated by 4mins of recovery in either conventional insoles or CSOs. Once completed the insole type was changed and the sprint protocol was repeated. Crank and joint-specific powers were obtained from instrumented force cranks and inverse dynamics methods respectively. Results from the paired samples t-test show no significant difference on a group level. Single subject analyses using magnitude based inferences show subjects could be grouped based on response (positive=2, non-responders=4, negative=6). Post-hoc analysis of joint-specific powers revealed negative responders tended to demonstrate reduced ankle reduced ankle power and range of motion ( $F= 4.97$ ; d.f. 1, 9,  $p= 0.05$ ), ( $F= 7.52$ ; d.f. 1, 9,  $p= 0.02$ ). The results highlight the need for caution when considering orthotic interventions and confirms the importance of the dual role of the ankle plantar flexors in cycling.

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## Introduction

In cycling, power is predominantly generated by the muscles surrounding the hip, knee and ankle joints during the downstroke phase of a revolution. The hip and knee extensors are responsible for approximately 55% of propulsive power over a cycle (Raasch, Zajac, Ma, & Levine, 1997). It has been shown that as intensity increases, the magnitude and method of power production is altered. The absolute power produced across the hip, knee and ankle joints is increased and there are reductions in the relative contribution of knee extension in favour of increased knee flexion power during the upstroke phase. Relative contributions of hip and ankle power however remain constant throughout (Elmer, Barratt, Korff, & Martin, 2011).

Recent studies that have focussed on the degree of muscle activation achieved by the lower limbs during maximal cycling have concluded that the hip extensors are underused (Dorel, Guilhem, Couturier, & Hug, 2012), (Watier, Costes, & Moretto, 2013). These studies found that the knee extensors and ankle plantar flexors are maximally recruited when normalised against off-bike maximal voluntary contractions (MVC). In contrast, the hip extensors are underused, achieving 77% of maximal recruitment of the gluteus maximus (Dorel, Guilhem, Couturier, & Hug, 2012) and 33% of maximal off-bike hip torque production (Watier, Costes, & Moretto, 2013).

A possible reason for this marginal use involves the way in which power is delivered from the muscles to the pedal. Raasch, Zajac, Ma, & Levine (1997) reported that only 44% of power generated during the downstroke is delivered directly to the pedal whilst the remaining 56% of muscle power is delivered to

the limb segments, and then transferred to the crank by the ankle plantarflexors. For this mechanism to function correctly the plantarflexors must stiffen the ankle joint sufficiently so that the power delivered to the limb can be effectively transferred to the pedal. Without this, the power produced by the hip and knee extensors would simply accelerate the limbs (cause hyperextension of the knee and dorsiflexion of the ankle) rather than generate pedal power (Raasch, Zajac, Ma, & Levine, 1997).

In support of this notion, a trend towards increased hip power has been seen when using rigid ankle braces to artificially stiffen the ankle joint (Barratt, 2015). This supports the notion that ankle strength is indeed a limiting factor of hip extension power in maximal cycling. However, although the braces provided extra support to the ankle joint, the reduction in the range of movement of the ankle resulted in a more than comparable decrease in ankle power and subsequently overall crank power. This highlights the need to strengthen the ankle joint without restricting its range of movement, and demonstrates the dual role of the muscles surrounding the ankle joint; to directly contribute to power production, and also to stiffen the ankle joint to transfer power to the pedal.

The prescription of orthotics for the prevention of injuries and enhancement of performance in cycling is well established (O'Neill, Graham, Moresi, Perry, & Kuah, 2011), (Yeo & Bonanno, 2014). There are few studies however that have investigated the performance enhancing capabilities of orthotics (Dinsdale & Williams, 2010), (Koch, Fröhlich, Emrich, & Urhausen, 2013), (Schmidt, Klaus, & Roth, 2011). These studies have centred on the notion that supporting the medial-longitudinal arch will reduce pronation, resulting in improved power



transfer from the lower limbs to the pedal. This is logical as the arch support helps to maintain the neutral position of the talonavicular joint providing a sturdy base from which the ankle can act. In theory, this would result in increased hip power through a strengthened ankle joint and increased overall power as the range of motion of the ankle is not compromised.

At present, the existing literature is inconclusive with regards to the effects of cycling specific orthotics (CSO) on cycling performance. Improvements of 6.9% in average sprint performance (Schmidt, Klaus, & Roth, 2011) and no significant difference in mean and peak power (Koch, Fröhlich, Emrich, & Urhausen, 2013), (Yeo, Rouffet, & Bonanno, 2015) have been reported when comparing CSOs to sham insoles. Two of the studies however possess major methodological limitations. Schmidt, Klaus, & Roth (2011) took their measurements two weeks apart in a non-randomised order. In addition, there was no deception used to reduce the confounding effects of a placebo. Conversely, Koch, Fröhlich, Emrich, & Urhausen (2013) randomised their trials and used a sham device. The main limitation to their study however was the use of consecutive Wingate Anaerobic Tests (WAnTs) which has been shown to induce fatigue thus affecting the accuracy and validity of the results (Yeo & Bonanno, 2014). Furthermore, the ergometer used in both studies (Cyclus 2) has not been mechanically validated (Bertucci, Grappe, & Crequy, 2011) and has thus not been shown to provide a valid or reliable measure of power (Yeo & Bonanno, 2014).

Despite the differences in study design, another potential explanation for the contrasting findings in the existing literature could involve the highly

individualistic responses to orthotics. O'Neill, Graham, Moresi, Perry, & Kuah, (2011) found no group effect when investigating kinematic responses to the use of orthotics. However, significant effects were seen on an individual level. Similarly, responses (mean power difference) with and without forefoot wedges has been found to correlate with degree of forefoot varus (Dinsdale & Williams, 2010). However due to the small sample size ( $n=6$ ), further investigation is required. Taken together, these results suggest that responses to orthotic interventions are highly dependent upon subject-specific foot morphology and function.

There are many ways to assess the morphology and function of the foot. One such method to be validated against radiographic data is the measurement of navicular height (Williams & McClay, 2000). More specifically, normalised navicular height using the total length of the foot as reference. In addition to establishing the height of the medial-longitudinal arch, the mobility of the foot can be described by taking measurements in both weight-bearing and non-weight-bearing conditions. It has been proposed that a large difference in the measurements taken during 10% weight-bearing and 90% weight-bearing describes a flexible, mobile foot whilst small differences in the measurements denotes a rigid foot (Williams & McClay, 2000). A good level of agreement has been reported between normalised navicular height and radiographic measurements (10% weight-bearing ICC= 0.914, 90% weight-bearing ICC= 0.924) (Williams & McClay, 2000). It is anticipated that this type of foot characterisation can predict subject-specific responses to orthotic use.

Furthermore, as seen in the existing literature, there is a need for a single-subject statistical approach when investigating this topic to minimise the chances of committing a type II error (O'Neill, Graham, Moresi, Perry, & Kuah, 2011), (Dinsdale & Williams, 2010). This approach has been encouraged in the analysis of movement and motor control but is infrequently used. It is thought that in many of these situations, highly individualised responses are more likely to be the rule than the exception (Bates, 1996). In addition, it has been argued that in studies investigating the effects of an intervention on performance, it is not only important to know whether the differences in the measurements are statistically significant but whether they are also meaningful to performance (Batterham & Hopkins, 2006). With this additional information, it may be possible to improve the prescription of orthotic interventions.

The aims of the present study therefore are twofold: 1) Assess whether cycling-specific orthotics improve maximal cycling performance and 2) Assess whether subject-specific responses correlate with foot characterisation. It is hypothesised that maximal cycling performance will improve with the use of CSOs and that subjects with mobile feet will experience greater levels of difference between conditions. It is also anticipated that the difference in maximal cycling performance will be a result of increased hip extension power.

## Methods

### Participants

Twelve recreational cyclists volunteered to take part in the study: eight male (age,  $38 \pm 8$  yr; height,  $180.41 \pm 3.55$  cm; body mass,  $80.90 \pm 6.50$  kg) and four female (age,  $35.92 \pm 20.82$  yr; height,  $176.57 \pm 1.94$  cm; body mass,  $77.20 \pm 2.05$  kg). All participants received verbal and written explanations of the experimental protocol and were told that they could withdraw from the study at any time without consequence. Once a health screen had been completed, participants gave their written informed consent to take part in the study. The experimental procedures were approved by the Faculty of Life Sciences Research Ethics Committee at the University of Chester.

### Design

The present study adopted a repeated measures design to assess the effect of cycling-specific orthotics (CSO) on maximal cycling performance. No habituation sessions were incorporated as evidence suggests the effect of orthotics on joint dynamics is immediate (MacLean, Davis, & Hamill, 2008). Insole type was used as the independent variable and the order in which these conditions were presented to each subject was randomised. Crank and joint-specific powers were the dependent measures obtained from instrumented force cranks and inverse dynamics methods respectively.

### Procedures

Participants visited the laboratory on one occasion. Upon arrival, subjects' height and body weight were recorded. Body weight was then used to calculate acceptable ranges for weight bearing and non-weight bearing navicular height

measurements. Subjects stood with their feet apart, left foot planted on a set of scales and right foot on the adjacent surface. They were then asked to distribute their weight onto their left foot without leaning to one side for the 10% weight bearing condition. This process was mirrored for the 90% weight bearing condition. The navicular tuberosity was found through palpation from the medial malleolus. Navicular height was then measured in weight bearing and non-weight bearing conditions for the right foot and normalised for foot length.

Once all anthropometric measurements were recorded, a total of five reflective markers were placed on the subjects to give an indication of hip, knee and ankle joint centres. Marker placement was as follows: Iliac crest (reference), greater trochanter (hip), lateral femoral condyle (knee), lateral malleolus (ankle) and pedal spindle. The markers were placed by the same investigator for each participant.

Subjects completed a 10 minute submaximal warm up at a self-selected pace followed by a single practice sprint (4 s) on a modified SRM cycling ergometer (SRM Ergometer, Schoberer Rad Messtechnik, Jülich, Germany).

Measurements of the participant's saddle height defined as the distance from the centre of the saddle to the centre of the bottom bracket; and handle bar reach defined as the distance from the tip/nose of the saddle to the centre of the handle bar were recorded and transferred to the ergometer. Crank length for all participants was set to 170mm and standardised cycling shoes (Specialized, Comp Road) were provided to minimize the effect of shoe design on the results.

The sprint protocol itself consisted of four short (4 s) all-out isokinetic sprints separated by recovery periods of 4 minutes each in agreement with Barratt (2015). Pedalling rate was held constant at 120 rpm by the ergometer. All sprints were initiated verbally by the researcher and subjects were given verbal encouragement for the duration of the sprint. Crank force data was collected by instrumented force cranks (Vector Cranks, BF1 Systems, Diss, UK) at a sampling rate of 200 Hz. The cranks measured force perpendicular to the pedal in addition to tangential force and crank angle. Joint specific power output was calculated from position data collected by a high speed camera (EX-F1, Casio, US) operating at 100 Hz. The camera was placed perpendicular to the sagittal plane of motion of the right leg. Once the motion files had been calibrated using an object of known horizontal and vertical distance, the position of the reflective markers in two dimensional space were quantified using the automatic digitisation process (Quintic Biomechanics v21, Coventry, UK).

### Data Processing

To reduce the signal-to-noise ratio, the data were filtered using a Butterworth low-pass filter. The cut-off frequency was selected using residual analysis (see appendix). The frequency at which there was a sharp increase in the gradient of the trend line of the residual plot was used as the cut-off frequency. This was seen to be 4Hz. Both kinematic and kinetic data were filtered at the same cut-off frequency as evidence indicates that failing to do so could incur large errors in the inverse dynamics calculations (Kristianslund, Krosshaug, & van den Bogert, 2012). Marker and force data were then input into a custom Microsoft Excel Spreadsheet set up to calculate joint specific power outputs using inverse dynamics methods.

## Statistical Analyses

All statistical analyses were performed using IBM SPSS Statistics for Windows (Version 21.0. Armonk, NY: IBM Corp) and Microsoft Excel. Due to the subject-specific nature of the study, a combination of group and individual statistics were used in agreement with Bates (1996). Firstly, to establish if a difference in mean crank power existed between the Control and CSO conditions on a group level, a paired samples t-test was used. Individual statistics were then performed by treating each condition as an independent sample in a t-test to identify responders and non-responders. For both sets of difference tests, parametric assumptions were met and the two-tailed version was adopted. The p value and t statistic for each independent samples t-test was then used to calculate the probabilities of where the true value of that t statistic lay (See Appendix). These were calculated based on thresholds of meaningful change of 2.5% improvement to be considered beneficial to performance and -0.5% decrement to be considered harmful to performance (Batterham & Hopkins, 2006). The threshold of 2.5% was chosen based on the group average crank power and was considered to be the smallest meaningful difference in performance. Finally, a linear regression was used to indicate to what extent the foot morphology and function of the subjects explained the variance between the conditions. The p-value and confidence limits for all tests was set at  $p < 0.05$  and 90% respectively.

## Results

Mean crank power across the 12 subjects for the control condition was  $444.5 \pm 119.5$  W compared to  $445.9 \pm 127.2$  W for the CSO condition. These values were not significantly different from one another ( $t = -0.27$ ; d.f. 11,  $p = 0.80$ ). Single-subject statistics show a non-significant result on an individual level (Table 1).

*Table 1 Descriptive statistics and null hypothesis tests on individual and group level for differences in crank power*

Subject	Control (W)	CSO (W)	Mean Difference (W)	t	P value
1	$584.9 \pm 14.5$	$581.8 \pm 10.2$	-3.14	-0.251	0.825
2	$396.7 \pm 7.3$	$422.1 \pm 9.5$	25.33	2.997	0.096
3	$425.2 \pm 55.8$	$410.7 \pm 14.0$	-14.50	-0.356	0.756
4	$402.7 \pm 0$	$381.4 \pm 48.9$	-21.33	-0.356	0.782
5	$222.2 \pm 5.7$	$209.1 \pm 22.2$	-13.04	-0.807	0.504
6	$541.1 \pm 0$	$563.9 \pm 12.6$	22.78	1.473	0.38
7	$518.1 \pm 3.8$	$502.7 \pm 7.2$	-15.43	-2.692	0.115
8	$658.5 \pm 0$	$662.5 \pm 31.5$	3.99	0.104	0.934
9	$404.6 \pm 13.7$	$393.0 \pm 0$	-11.54	-0.688	0.617
10	$349.3 \pm 1.4$	$390.0 \pm 21.6$	40.68	2.656	0.117
11	$305.0 \pm 12.7$	$305.5 \pm 0.6$	0.51	0.056	0.96
12	$525.1 \pm 23.6$	$528.4 \pm 2.6$	3.30	0.197	0.862
Group	$444.5 \pm 119.5$	$445.9 \pm 127.2$	1.47	-0.265	0.796

Magnitude based inferential statistics reveal two possible responders in subject 2 and subject 10 (Table 2). The probability that the true value of the t statistic is positive and meaningful is 76.4% for subject 2 and 59.7% for subject 10 (Figure 1). Six negative responders were also found during this process with an average probability of 57.0% that the t statistic is negative and meaningful (Figure 1).



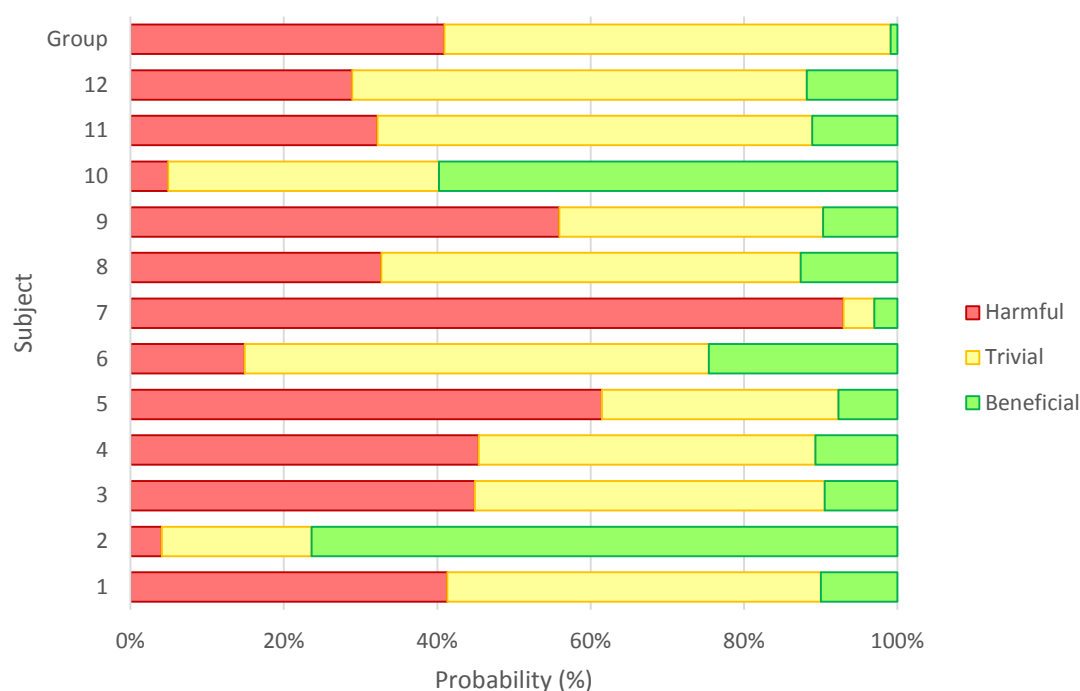


Figure 1 Probabilities that intervention is harmful, trivial or beneficial to performance based on thresholds of 2.5% change for beneficial and -0.5% change for harmful

Table 2 Clinical and mechanistic inferences based on probabilities (Figure 1)

Subject	Clinical Inference	Mechanistic Inference
1	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
2	Likely beneficial; Very unlikely harmful; Use	Likely positive
3	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
4	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
5	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
6	Unlikely harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
7	Likely harmful; Very unlikely beneficial; Do not use	Likely negative
8	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
9	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
10	Unclear; Do not use; Collect more data	Possibly positive
11	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
12	Possibly harmful; Unlikely beneficial; Do not use	Unclear; Collect more data
Group	Possibly harmful; Most unlikely beneficial; Do not use	Possibly trivial

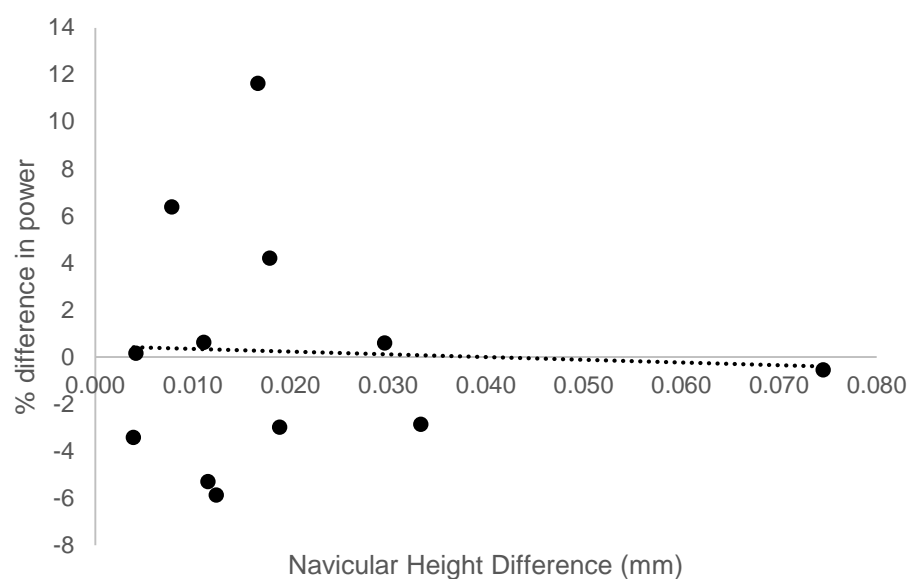
To assess the impact of the CSOs on hip extension power, group and single-subject statistical tests (paired and independent samples t-tests) were conducted. Due to issues in the synchronisation of kinematic and kinetic data, subject 4 was excluded from analysis involving inverse dynamics methods. The

results (Table 3) show no significant difference in hip extension power between conditions ( $p>0.05$ ).

*Table 3 Descriptive statistics and null hypothesis tests on individual and group level for differences in hip extension power*

Subject	Control (W)	CSO (W)	t	P value
1	319.25 $\pm$ 5.50	333.89 $\pm$ 51.56	-0.399	0.728
2	310.39 $\pm$ 24.69	317.52 $\pm$ 19.84	-0.318	0.781
3	213.00 $\pm$ 29.41	212.72 $\pm$ 60.44	0.006	0.996
5	229.75 $\pm$ 36.09	136.33 $\pm$ 208.33	0.625	0.596
6	345.65 $\pm$ 0	343.57 $\pm$ 6.12	0.277	0.828
7	359.88 $\pm$ 10.82	353.91 $\pm$ 22.15	0.342	0.765
8	492.28 $\pm$ 0	466.08 $\pm$ 57.69	0.371	0.774
9	210.39 $\pm$ 9.33	235.60 $\pm$ 0	-2.204	0.271
10	218.60 $\pm$ 9.72	234.39 $\pm$ 15.87	-1.2	0.353
11	211.77 $\pm$ 29.81	235.30 $\pm$ 13.73	-1.014	0.417
12	348.81 $\pm$ 4.73	403.39 $\pm$ 26.29	-2.889	0.102
Group	288.31 $\pm$ 79.58	287.56 $\pm$ 101.88	0.049	0.961

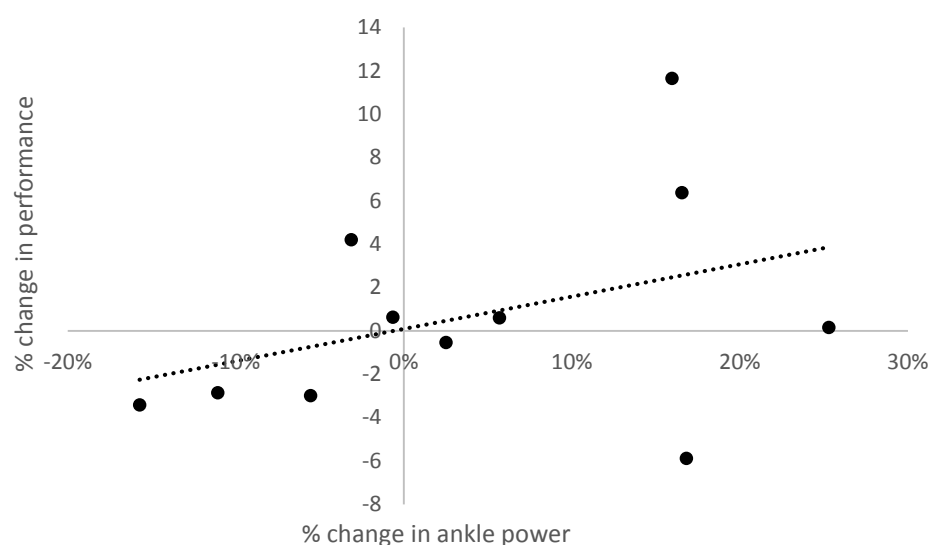
A linear regression was calculated (Figure 2) to predict percentage change in performance based on the difference in normalised navicular height scores (foot mobility). A non-significant regression equation was found ( $F= 0.01$ ; d.f. 1, 11,  $p= 0.95$ ).



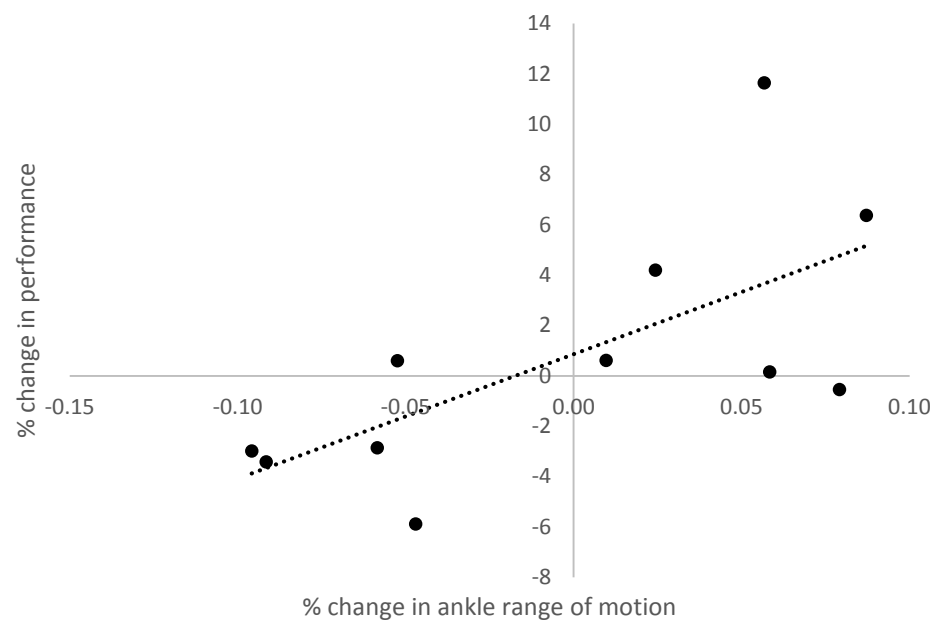
*Figure 2 Difference in navicular height measures at 10% and 90% weight bearing as a predictor of performance in CSO condition*

As the results did not support the hypothesis that CSO would increase crank power due to increased hip extension power, post hoc analyses of other joint specific powers were conducted to better understand the factors contributing to the individual positive and negative responses observed. Specifically, an analysis of ankle power and ankle range of motion was undertaken due to the possible effects of the CSO on the ability of the ankle joint to produce power; the additional role of the ankle joint alongside transferring limb power.

Linear regression results identified a significant correlation between individual responses to CSO (% change in performance) and both % change in ankle power and % change in ankle range of motion ( $F= 4.97$ ; d.f. 1, 9,  $p= 0.05$ ), ( $F= 7.52$ ; d.f. 1, 9,  $p= 0.02$ ).



*Figure 3 Trend shows a lesser positive relationship between ankle power and maximal cycling performance.*



*Figure 4 Trend shows a positive relationship between ankle range of motion and maximal cycling performance.*

## Discussion

This study was designed to determine the effects of cycling-specific orthotics on maximal cycling performance. Results of the paired samples t-test show no significant difference between mean crank power values for control and CSO conditions on a group level ( $t = -0.27$ ; d.f. 11,  $p = 0.80$ ). However, the use of magnitude-based inferential statistics identified two likely positive responders, four non-responders and six likely negative responders. Positive responses to CSO were not due to increased hip extension power as hypothesised and there was no significant association between foot mobility and individual responses. It was found that the power-producing capability of the ankle joint in each of the conditions could explain the observed responses. This could be of great practical importance as the two positive responders demonstrated increases of 5% and above in crank power. These values would be considered extremely meaningful to a cyclist and would require months/years of strength training and practice to obtain. Conversely, four of the negative responders produced decreases in crank power of more than 2%. This could also be considered meaningful and highlights the need for coaches and athletes to be cautious when considering orthotic interventions.

Regarding the first aim of this study, to determine the effects of CSO on cycling performance, both group and single-subject analysis using traditional null hypothesis tests showed no significant difference in crank power between conditions. This is consistent with previous studies that have reported null findings (Koch, Fröhlich, Emrich, & Urhausen, 2013), (Yeo, Rouffet, & Bonanno, 2015) and opposes that of Schmidt, Klaus, & Roth (2011). One possible reason for the difference in results when compared to Schmidt, Klaus, & Roth (2011) is

that, in both the present study and that of Koch, Fröhlich, Emrich, & Urhausen (2013), the orthotics were not tailored to each individual. In the present study foot characterisation was used in an attempt to account for this methodological difference by scaling the individual's response to their foot type. It is also possible that tailoring the level of support to the individual would have had little to no effect on the outcome as Yeo, Rouffet, & Bonanno (2015) also reported a null finding when employing custom-made orthotics.

This finding was expanded upon when exploring the data using a reference value considered to be the smallest meaningful (positive) change in performance (2.5%). It was revealed that two of the twelve subjects were likely to benefit from the intervention shown by 79.4% and 59.7% probability of improved performance. In contrast, six subjects displayed negative responses to the CSO based on the smallest meaningful (negative) change in performance of 0.5%. This supports the notion that responses to orthotic interventions are highly specific to the individual. This was also seen in past studies investigating the effects of orthotics (Dinsdale & Williams, 2010), (O'Neill, Graham, Moresi, Perry, & Kuah, 2011). The magnitude based inferential statistics also highlighted the need for additional trials, in order to make clear inferences for many of the subjects. Future studies adopting a single-subject approach should be aware of this and plan accordingly.

The second aim of the study was to assess whether an individual's response to the CSOs related to their foot characterisation, and evaluate its predictive power. The results of the linear regression ( $F = 0.01$ ; d.f. 1, 11,  $p = 0.95$ ) suggest there is poor predictive power in navicular height difference (foot mobility) when

investigating maximal power output. This is contrary to the notion that individuals with higher levels of foot malalignment (over-pronation due to mobile feet) demonstrate larger responses to posture correcting orthotics (Dinsdale & Williams, 2010). A possible explanation for this difference is that Dinsdale & Williams (2010) looked specifically at the effects of forefoot wedges on varying levels of forefoot varus. Forefoot wedges occupy a different position under the foot and are not as firm as the orthotics used in this study. An argument could therefore be made that comfort plays a large role in an individual's response to orthotic interventions. Also, the number of wedges used were calculated based on each individual's forefoot varus score while there was no tailoring used in the present study.

Following analysis of the magnitude based inferential statistics, where it was revealed that there were likely two positive responders, it was seen that subjects could be grouped by response (positive= 2, non-responders=4, negative=6). Post-hoc analysis of joint-specific powers and kinematic values revealed that negative responders tended to demonstrate reduced ankle range of motion, and potentially as a result, reduced ankle power ( $F= 7.52$ ; d.f. 1, 9,  $p= 0.02$ ), ( $F= 4.97$ ; d.f. 1, 9,  $p= 0.05$ ). This is logical as reducing the dorsiflexion range of motion also decreases the plantar flexor moment arm resulting in a smaller plantar flexor moment for a given muscle force. This concept is supported by previous research investigating walking gait (Mueller, Minor, Schaaf, Strube, & Sahrmann, 1995) and contradicts conventional wisdom that ankle movement is of little importance to cycling. It also further supports the theory that ankle range of motion should not be compromised to increase ankle strength. Once again highlighting the importance of the dual role of the ankle

plantar flexors to not only stiffen the ankle sufficiently to deliver the power from the limbs to the crank, but also to contribute to crank power. This secondary role cannot be achieved without movement and would result in a loss of ankle power which has been found to make up approximately 11% of crank power during maximal cycling (Barratt, 2015).

The results of the present study taken together with those of Koch, Fröhlich, Emrich, & Urhausen (2013) and Yeo, Rouffet, & Bonanno (2015) suggest that foot posture, and attempts to correct it, have no significant effect on performance. As aforementioned, comfort may play a large role in an individual's response to foot orthoses and may explain the variance in this study. However, previous studies that have incorporated perceived comfort rating systems have found no significant difference between conventional insoles and custom-made orthotics (Bousie, Blanch, McPoil, & Vicenzino, 2013), (Yeo, Rouffet, & Bonanno, 2015). It is thought that the definition of comfort is too vague and therefore difficult to measure accurately (Nigg, Nurse, & Stefanyshyn, 1999). A more appropriate variable to quantify would be proprioception. A concept presented previously in the literature, may better explain the results of the present study. Nigg, Nurse, & Stefanyshyn (1999) considered orthotics to be just one level of a filter, from the reaction force signal to mechanoreceptors in the foot. It is thought that this filtered information is relayed to the central nervous system (CNS) where a dynamic response is selected on the basis of individual constraints. In addition, the sensitivity of the mechanoreceptors in the foot varies between individuals, with subject specific thresholds. Finally, it was suggested that subjects with similar sensory thresholds demonstrate similar movement responses. This could explain why



there were three clear groups (positive, non-responders, negative) and why ankle range of motion was increased in the positive responders and decreased in the negative responders. This mechanism cannot be confirmed however as subject specific sensory thresholds were not explored.

It is thought that preferred movement paths exist for any given task (Nigg, 2001). This movement path is again selected based on individual constraints and deviations from this path result in increased muscular activity. Therefore it seems logical to assume that the higher the quality of information relayed to the CNS, the more accurate and efficient the dynamic response. Research into the effects of increased proprioceptive feedback on movement has mostly adopted textured insoles. These insoles have been seen to alter muscular activity (Murley, Landorf, Menz, & Bird, 2009) and improve movement discrimination (Waddington & Adams, 2000). However, the research around this topic has been conducted with injury prevention in mind, and to the author's knowledge, there are no studies investigating the effects of textured insoles on sporting performance. Given the findings of this study, it seems possible that sprint cycling performance could be influenced by the proprioceptive feedback of the feet. Future research utilising textured insoles may therefore provide further insight into the role of proprioception during sprint cycling and may even be an avenue for performance enhancement.

A potential limitation of the present study involves the choice and collection of navicular height at 10% and 90% weight bearing as a descriptor of foot mobility. Although this method has been shown to be valid and reliable, with reference to values derived from radiography, it is not considered the gold standard for

quantifying foot posture (Williams & McClay, 2000). Also, as navicular height difference has not been used in cycling, there is no normative data to compare the current findings to. Finally, the measurements taken in this study are not analogous with previous work in clinical settings which highlights the possibility of errors made during data collection. The lack of accurate foot posture data makes it difficult to draw definitive conclusions about the effect of posture correcting orthotics on maximal cycling performance on the basis of this study alone. However the crank power results support the findings of Yeo, Rouffet, & Bonanno (2015) who recruited participants with high levels of foot mobility as they were thought to benefit most from the intervention.

## **Conclusion**

On average the CSO did not significantly improve maximal cycling performance or hip extension power. They did however produce highly individualistic responses with regards to cycling performance, many of which were practically meaningful in magnitude. The CSO produced both positive and negative responses emphasising the importance of caution when considering orthotic interventions of this nature. These responses were explained by changes in ankle power and ankle range of motion due to the CSO. This confirms the importance of the dual role of the ankle joint in cycling.

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## Appendix 1



### Participant information sheet

#### **The effect of increasing foot rigidity on maximal cycling power through the use of cycling specific orthotics**

You are being invited to take part in a research study. Before you decide, it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

Thank you for reading this.

#### **What is the purpose of the study?**

The main aim of the study is to investigate how a pair of custom cycling orthotics effect all-out sprint cycling performance.

#### **Why have I been chosen?**

You have been invited because you are trained in the sport being investigated, minimizing the learning effect associated with the testing protocol. Also, as a competitive cyclist any findings from the study may be applicable to you.

#### **Do I have to take part?**

It is up to you to decide whether or not to take part. If you decide to take part you will be given this information sheet to keep and be asked to sign a consent form. If you decide to take part you are still free to withdraw at any time and without giving a reason. A decision to withdraw at any time, or a decision not to take part, will not affect you in any way.

#### **What will happen to me if I take part?**

The study will take place over the summer. You will be asked to visit the laboratory on one occasion with the session lasting approximately 2hrs. This session will consist of an initial static foot measurement to characterize your feet, reflective markers placed on your hips, knees and ankles, followed by a warm up of 10 minutes of submaximal cycling on a cycling ergometer and a single sprint of 4 seconds. You will then be asked to perform 2 sprints lasting 4 seconds each with 4 minutes rest between them. This will be done in one of three conditions (standard insoles, custom cycling orthotics, textured insoles). Once the first 2

sprints have been completed, you will be asked to complete 2 more sprints in the same fashion for the other conditions (6 total sprints). During this time you will be filmed and the crank power recorded to estimate the power contribution of each of your joints.

**What are the possible disadvantages and risks of taking part?**

The test requires all-out effort during the 4s sprint so there is the risk of discomfort and muscle soreness predominantly in your legs. Also, if you are unfamiliar with orthotics, they may feel uncomfortable at first.

**What are the possible benefits of taking part?**

By taking part in the study you will have the opportunity to learn more about the characteristics of your feet and how this impacts on performance and injury in cycling. You will also find out if realignment of the foot through the use of orthotics may work for you.

**What if something goes wrong?**

If you wish to complain or have any concerns about any aspect of the way you have been approached or treated during the course of this study, please contact Dean of the Faculty of Life Sciences, University of Chester, Parkgate Road, Chester, CH1 4BJ, 01244 513055.

**Will my taking part in the study be kept confidential?**

All information which is collected about you during the course of the research will be kept strictly confidential so that only the researcher carrying out the research will have access to such information.

**What will happen to the results of the research study?**

The results will be written up into a dissertation for my final project of my MSc. Individuals who participate will not be identified in any subsequent report or publication.

**Who is organising the research?**

The research is conducted as part of an MSc in Sports Biomechanics within the Department of Sport and Exercise Sciences at the University of Chester. The study is organised with supervision from the department, by Luke Sharland-Wong, an MSc student.

**Who may I contact for further information?**

If you would like more information about the research before you decide whether or not you would be willing to take part, please contact:

**Thank you for your interest in this research.**

## Appendix 2



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Chester



## Pre-test Questionnaire

**The effect of increasing foot rigidity on maximal cycling power through the use of cycling specific orthotics**

**Researcher:**

Name: \_\_\_\_\_ Test date: \_\_\_\_\_

Contact number: \_\_\_\_\_ Date of birth: \_\_\_\_\_

In order to ensure that this study is as safe and accurate as possible, it is important that each potential participant is screened for any factors that may influence the study. Please circle your answer to the following questions:

1. Has your doctor ever said that you have a heart condition *and* that you should only perform physical activity recommended by a doctor? YES/NO
2. Do you feel pain in the chest when you perform physical activity? YES/NO
3. In the past month, have you had chest pain when you were not performing physical activity? YES/NO
4. Do you lose your balance because of dizziness *or* do you ever lose consciousness? YES/NO
5. Do you have bone or joint problems (e.g. back, knee or hip) that could be made worse by a change in your physical activity? YES/NO
6. Is your doctor currently prescribing drugs for your blood pressure or heart condition? YES/NO
7. Are you pregnant, or have you been pregnant in the last six months? YES/NO
8. Have you injured your hip, knee or ankle joint in the last six months? YES/NO
9. Do you know of any other reason why you should not participate in physical activity? YES/NO

Thank you for taking your time to fill in this form. If you have answered 'yes' to any of the above questions, unfortunately you will not be able to participate in this study.



### Appendix 3



University of  
Chester



**Title of Project: The effect of increasing foot rigidity on maximal cycling power through the use of cycling specific orthotics**

**Name of Researcher:**

**Please initial box**

1. I confirm that I have read and understand the information sheet for the above study and have had the opportunity to ask questions. ☐
2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason and without my legal rights being affected. ☐
3. I agree to take part in the above study. ☐

\_\_\_\_\_  
Name of Participant

\_\_\_\_\_  
Date

\_\_\_\_\_  
Signature

\_\_\_\_\_  
Researcher

\_\_\_\_\_  
Date

\_\_\_\_\_  
Signature

1 for participant; 1 for researcher